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The Effect of Dynamic Hip Motion on the Micromotion of Press-Fit Acetabular Cups in Six Degrees of Freedom

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Abstract

The hip joint is subjected to cyclic loading and motion during activities of daily living and this can induce micromotions at the bone-implant interface of cementless total hip replacements. Initial stability has been identified as a crucial factor to achieve osseointegration and long-term survival. Whilst fixation of femoral stems achieves good clinical results, the fixation of acetabular components remains a challenge. *In vitro* methods assessing cup stability keep the hip joint in a fixed position, overlooking the effect of hip motion. The effect of hip motion on cup micromotion using a hip motion simulator replicating hip flexion-extension and a six degrees of freedom measurement system was investigated. The results show an increase in cup micromotion under dynamic hip motion compared to Static Flexion. This highlights the need to incorporate hip motion and measure all degrees of freedom when assessing cup micromotion. In addition, comparison of two press-fit acetabular cups with different surface coatings suggested similar stability between the two cups. This new method provides a basis for a more representative protocol for future pre-clinical evaluation of different cup designs.

Keywords

Total hip replacement, acetabular cups, press-fit fixation, initial stability, biomechanical study

Abbreviations

CNC = Computer numerical control

DoF = Degrees of Freedom

HA = hydroxyapatite

LVDT = Linear Variable Differential Transformer

THR = Total Hip Replacements

Introduction

The hip joint is subjected to cyclic loading and motion during activities of daily living, which can induce micromotion of total hip replacements (THR) [1,2]. High levels of micromotion can inhibit bone formation and hence osseointegration of cementless implants. Studies have reported that osseointegration of cementless implants will occur if the relative micromotion at the bone-implant interface is below 40 μm and may occur up to 150 μm [3,4]. Micromotions above 150 μm can result in fibrous tissue formation, which contributes to implant loosening and lead to a requirement for revision surgery [5]. Initial implant stability is therefore crucial for the long-term survival of implants [6] and it is important to be able to measure implant micromotion in the pre-clinical testing of new implants and fixation philosophies. This is especially true for the acetabular component as cup fixation can be challenging.

Even though there is a clear link between initial stability and osseointegration of the acetabular cup, there is no standardised pre-clinical test protocol available to investigate the levels of micromotion, and hence initial stability, of cementless press-fit acetabular cups. There are, however, numerous studies available in the literature that measure the micromotion of acetabular cups implanted into cadaveric bones [2,7–11]. These studies have two main limitations: firstly, none have investigated the motion of acetabular cups in all six degrees of freedom (DoF); instead they only reported one or a few selected directions of motion assumed to be dominant. Secondly, the hip is set at a fixed angle throughout testing, usually replicating heel strike or single leg stance; however, no studies have taken into account the effect of dynamic hip motion as seen during activities of daily living on cup micromotion.

A previous study tackled the first limitation with a system capable of measuring the micromotion of a cup in six DoF when it was subjected to cyclic loading (Figure 1)[12]. The results from this study showed significant levels of micromotion in different directions of motion, highlighting the importance of measuring cup micromotion in all six directions of motion. However, the effect of dynamic hip motion was not investigated in that study; instead, the hip was held at a fixed position modelling single leg stance. In addition, the previous study also introduced a synthetic acetabular model using Sawbones polyurethane foam blocks which replicated the important structural features of the acetabulum (Figure 1).

[Insert Figure 1]

The aim of this study was to consider the second limitation of micromotion studies by investigating the effect of dynamic hip motion on cup micromotion. This was carried out by combining both the six DoF measurement system and the acetabular model developed previously to a dynamic hip motion simulator capable of replicating hip flexion-extension as seen during activities of daily living. In addition, two press-fit acetabular cups with different porous coatings were tested to assess the effect of surface coatings on the micromotion of press-fit cups.

Materials and Methods

Two press-fit acetabular components were used in this study (Figure 2): a Trident acetabular cup with a hydroxyapatite (HA) coating and a Tritanium acetabular cup (both cups from Stryker, Mahwah, NJ, USA); and their corresponding polyethylene liners (X3 liners, Stryker). The external diameter of both cups was 54 mm and the liners were for a 28 mm diameter femoral head.

[Insert Figure 2]

The Tritanium cup is a similar design to the Trident cup and is coated with a 3D titanium matrix designed to resemble trabecular bone and aims to improve cup osseointegration [13]. The two acetabular components were chosen as they offer the unique opportunity to assess the effect of a new porous coating (the 3D titanium matrix) on cup micromotion whilst keeping other variables, such as geometry and fixation type, the same.

Thirty Sawbones polyurethane foam blocks (density of 0.48 g/cm^3, compressive strength of 18 MPa, Sawbones block #1522-04) were used as a synthetic bone substrate. These blocks have similar mechanical properties to trabecular bone [14,15] and are commonly used to model the acetabulum in biomechanical tests [6,16–19].

The cavity manufactured in each block was hemispherical, representing the acetabulum, with two rectangular cavities superiorly and inferiorly to the acetabulum, representing the radiolucent triangle and the acetabular notch, respectively (Figure 1). This acetabular model was designed to replicate the structural properties observed around the acetabulum, which comprised compressive forces anteriorly and posteriorly from the acetabular columns and non-supportive areas superiorly and inferiorly caused by the presence of the radiolucent triangle and the acetabular notch, respectively, resulting in a pinching effect. This model was used in a previous study [12], and others have similarly been used in studies to assess acetabular shell deformation in both THR and hip resurfacing implants [20–22], stress distribution within press-fitted ceramic liners [23], and cup turn-out due to joint frictional moments [24].

The cavities were manufactured using a computer numerical control (CNC) machine rather than a reamer to ensure accuracy and consistency of the cavities. The hemispherical cavities were machined to a 53 mm diameter to achieve a 1 mm press-fit. A 1 mm press-fit was chosen rather than a 2 mm press-fit to ensure that full seating of the cup was achieved. Furthermore, a few studies have reported increased stability of cups implanted with a 1 mm press-fit compared to a 2 mm press-fit [2,9,25]. Prior to testing, the peripheral diameters of each acetabular cavity were measured using a digitiser (Incise, Renishaw, Wotton-under-Edge, UK) to confirm the accuracy of the machining; the mean peripheral diameter was 52.99 ± 0.02 mm.

To measure the micromotion of the cup in six DoF, the system presented in Figure 1 was used. The target frame with three spheres was rigidly attached to the acetabular cup using a rod screwed into the dome screw hole of the cup. The three spheres were in contact with six linear variable differential transformers (LVDTs; Red Crown LVDT, Marposs, Bentivoglio, Italy), positioned in different locations and directions. This configuration provided information on the position of all three spheres at any given moment during a test. These measurements were then used to calculate the three translational ranges of micromotion of the cup in the orthogonal axes (X, Y and Z) and the rotation about these axes (θ_X, θ_Y and θ_Z) from a single point of attachment (Figure 3). Translations in X, Y and Z are equivalent to anterior-posterior, superior-inferior, and medial-lateral motions, respectively; and rotations in θ_X, θ_Y and θ_Z are equivalent to medial-lateral tilt, anterior-posterior tilt, and medial-lateral rotation, respectively. Due to the set-up of the six DoF measurement system relative to the acetabular cup the measured micromotions included some deformation of the Sawbones foam. This deformation would have been the same in both the static and dynamic motion loading studies and did not detract from the study aim which was to compare the effect of these two loading scenarios.

Both the Trident and the Tritanium cups were implanted with a 1 mm press-fit into the acetabular cavities using a protocol of five sinusoidal load cycles (between 0.01 kN and 5.0 kN), applied at a frequency of 1 Hz using a single-axis hydraulic machine (Dartec, Series HC10, Zwick Testing Machines Ltd, Leominster, UK) [12].

Once implanted into the Sawbones block and connected to the six DoF system, the acetabular cup was placed at 45° inclination on a dynamic hip motion simulator, which modelled hip flexion and extension as a sinusoidal wave. The dynamic hip motion simulator was placed on the single-axis hydraulic machine and the cup was loaded vertically through a 28 mm femoral head connected to the actuator of the single-axis hydraulic machine (Figure 3).

[Insert Figure 3]

Three different test conditions were assessed for a total of 500 steps. The first was Static Flexion where the hip was held in a fixed position at 30° flexion and the cup was cyclically loaded between 0.01 kN and 2.0 kN at a frequency of 1 Hz. This modelled heel strike, and represented the test conditions commonly used in micromotion studies in the literature. The second condition simulated Level Walking, where, in addition to cyclically loading the cup to a peak load of 2.0 kN at 1 Hz, the hip was also dynamically flexed and extended from 30° flexion to 10° extension at a frequency of 0.5 Hz. The loading and flexion-extension cycles were synchronised to achieve peak loads at both heel strike and toe-off. The third condition simulated Stair Climbing. The test protocol was similar to that of Level Walking; however, the angles of flexion-extension were modified to 40° flexion and 5° extension. This reflected the greater range of motion and more flexed position of the hip joint during Stair Climbing compared to Level Walking [26–28]. The chosen angles of flexion and extension for both Level Walking and Stair Climbing were in line with those published in the literature [29–32].

Each test condition (Static Flexion, Level Walking and Stair Climbing) was repeated five times for each acetabular cup (Trident cup with HA coating and Tritanium cup) with a new acetabular model every time, resulting in a total of thirty tests.

Following the micromotion test, the cup was removed from the Sawbones blocks and cleaned with a soft nylon brush to remove any foam debris incrusted into the porous coating before being implanted into a new Sawbones block for the following test. Using the same acetabular component for all tests was acceptable after a preliminary study showed no changes or trends in micromotion of the cup with repeated use under similar conditions.

Non-parametric tests with a type I error of $\alpha = 0.05$ were performed for statistical analysis as the sample size ($n = 5$) was too small to prove normality. The Freidman and the Wilcoxon signed ranks post hoc tests were used to assess any difference among translations and rotations within each groups. The Kruskal-Wallis and the Mann-Whitney post hoc tests were used to assess the differences in micromotion when comparing the different testing conditions and the acetabular cups. All statistical analyses were carried out using SPSS (IBM, New York, USA).

Results

Similar patterns in micromotion were observed regardless of the test condition and acetabular component (Figures 4 and 5). Considering that 0.08° in rotation corresponded to a 40 μm displacement in the direction of the arc for both cups, the translations were always greater than the rotations. Within the translations, X was greater than Y and Z; and in the case of the rotations, θ_Y was generally greater than θ_X and θ_Z.

Micromotion of the Trident Cup with HA Coating

The micromotion of the Trident cup with HA coating was generally greater under dynamic hip motion (both Level Walking and Stair Climbing) compared to Static Flexion (Figure 4). The micromotion of the cup was significantly greater in all translations and in θ_Y when subjected to dynamic hip motion compared to Static Flexion (Static Flexion vs. Level Walking: X: $p = 0.021$, Y: $p = 0.009$, Z: $p = 0.008$ and θ_Y: $p = 0.005$; Static Flexion vs. Stair Climbing: X: $p = 0.009$, Y: $p = 0.016$, Z: $p = 0.009$ and θ_Y: $p = 0.011$). The micromotion of the cup was also significantly greater in both X and Z under Stair Climbing compared to Level Walking ($p = 0.047$ and $p = 0.008$, respectively).

[Insert Figure 4]

Micromotion of the Tritanium Cup

The micromotion of the Tritanium cup was significantly greater in all six DoF when subjected to dynamic hip motion compared to Static Flexion (Figure 5; Static Flexion vs. Level Walking: X: $p = 0.009$, Y: $p = 0.009$, Z: $p = 0.009$, θ_X: $p = 0.031$, θ_Y: $p = 0.006$ and θ_Z: $p = 0.013$; Static Flexion vs. Stair Climbing: X: $p = 0.009$, Y: $p = 0.009$, Z: $p = 0.009$, θ_X: $p = 0.020$, θ_Y: $p = 0.005$ and θ_Z: $p = 0.013$). The micromotion of the cup was also significantly greater in θ_Y under Stair Climbing compared to Level Walking (p = 0.018).

[Insert Figure 5]

Comparing the Trident Cup with HA Coating and the Tritanium Cup

The data presented in Figures 4 and 5 were rearranged in order to compare the micromotion of Trident to that of the Tritanium cup for each test condition (Figures 6 to 8). When tested under Static Flexion, the only significant difference was the Z micromotion, which was greater with the Tritanium cup (p = 0.045). Under Level Walking, the X, θ_X and θ_Y micromotions of the Tritanium cup were significantly greater than those of the Trident cup (p = 0.047, p = 0.033 and p = 0.017, respectively). Finally, only θ_X was significantly greater (p = 0.032) with the Tritanium cup under Stair Climbing compared to the Trident cup.

[Insert Figure 6]

[Insert Figure 7]

[Insert Figure 8]

Discussion

The aim of this study was to assess the effect of dynamic hip motion and two different porous coatings on the micromotion of press-fit acetabular cups. This study used both the experimental system to measure the micromotion of the cup in six DoF under cyclic loading and the acetabular model developed in a previous study [12]; and combined them with a dynamic hip motion simulator that modelled hip flexion-extension. With this setup, the micromotion of two acetabular cups (a Trident cup with HA coating and a Tritanium cup) were assessed under three different conditions: Static Flexion, Level Walking and Stair Climbing. Static Flexion, where the hip joint was kept at 30° flexion throughout the micromotion test, simulated heel strike. This test condition provided a reference with which to compare the micromotion of the cups when subjected to dynamic hip motions. Level Walking and Stair Climbing were chosen as they are both common activities of daily living that patients undertake relatively soon following surgery, and therefore conditions that the implants must be able to withstand. The maximum load the hip joint is subjected to during Stair Climbing is slightly higher than that of Level Walking [29,31,33,34]. In this study, however, the loading protocol was kept constant for all test conditions to provide a direct interpretation of the differences between Stair Climbing, Level Walking and Static Flexion as a function of the direction in joint reaction force only.

Similar to the findings of a previous study [12], significant levels of micromotion were observed in all translations, highlighting the importance of measuring all six DoF and not assuming a dominant direction of motion. Furthermore, the dominant direction of motion assumed in published micromotion studies is usually in the direction of cup insertion into the acetabulum, which is the equivalent to the Z micromotion in this study [2,7–11,35]. The results from this study, however, clearly show that the micromotions in X and Y were either similar or greater than in Z for all test conditions (Figures 4 and 5).

There was a significant increase in the micromotion of the cup when it was subjected to both Level Walking and Stair Climbing compared to Static Flexion, regardless of the cup tested (Figures 4 and 5). This observation was also reported in a study investigating the effect of dynamic hip motion on femoral stems [36]. An increase in micromotion of the cup when subjected to dynamic hip motion was expected as the direction of the joint reaction force was no longer static, but varied throughout the motion. This variation in direction of the joint reaction force can generate toggling motion of the cup, resulting in an increase in cup micromotion in all directions.

The micromotion of the cup also tended to be greater when it was subjected to Stair Climbing compared to Level Walking (Figures 4 and 5). An increase in micromotion with Stair Climbing compared to Level Walking was expected. The range of motion of the hip was increased during Stair Climbing compared to Level Walking (50° and 45°, respectively); however the frequency at which it was operated remained the same (0.5 Hz). This resulted in a more rapid variation in the change in direction of the joint reaction force, and hence, in an increase in cup micromotion. The Trident cup was more affected by the increase in range of motion between Level Walking and Stair Climbing compared to the Tritanium cup. Indeed, the increase in cup micromotion was significant in both X and Z with the Trident cup; these translations were both above 40 μm and therefore could have an influence on osseointegration of the cup. On the other hand, only θ_Y, which was below 40 μm, was significantly greater with the Tritanium cup. As there was no significant increase in micromotion between Level Walking and Stair Climbing amongst the clinically relevant motions (X, Y, and Z) with the Tritanium cup, it can be hypothesised that changes in range of hip motion had little effect on the stability of this cup.

This study also aimed to assess the micromotion of two cups with different porous coatings. These were the Trident cup with HA coating and the Tritanium cup. Only a few significant differences in micromotion were observed between the Trident cup and the Tritanium cup. The Tritanium cup exhibited significantly higher levels of micromotion in Z during Static Flexion (Figure 6); in X, θ_X and θ_Y during Level Walking (Figure 7) and in θ_X during Stair Climbing (Figure 8). Most of these significant differences were below 40 μm, which would have no detrimental effect on cup osseointegration. Hence, due to the lack of overwhelming differences in results between both cups, one can argue that the new Tritanium cup exhibits a similar level of stability as the Trident cup.

Comparing the micromotions obtained in this study to those in the literature is challenging as there are a limited number of studies available. Also, each has been carried out under specific test conditions, making them difficult to compare with each other. Hence, all the studies assessing micromotion of only press-fit cups were considered. The main differences in testing protocols that could affect cup micromotion were: level of press-fit, peak load, and orientation of the hip or the cup (Table 1). The micromotions in these studies were compared to those obtained during Static Flexion in this study. Furthermore, in order to limit the effect of load, the micromotions from each study were normalised with respect to the applied load prior to comparison (Figure 9).

[Insert Table 1]

[Insert Figure 9]

In general, the micromotions under Static Flexion reported in this study were similar to those reported in the literature when comparing micromotions in similar directions. Differences in micromotion of similar direction between different studies can primarily be attributed to the variation in cup positioning for testing. Indeed, the direction in which the cup is loaded will dictate the direction in which it will move. Comparing the micromotions with the Trident cup in this study, and those in Crosnier et al. [12], the only difference in the test protocol was the orientation of the cup when loaded (heel strike in this study and single leg stance in the previous one). In this case, the micromotion in Z went from being the greatest (previous study) to the smallest (this study) of the three translations.

As discussed, the micromotion measurements include the deformation of the Sawbones block however this effect was likely to be mostly in the Z direction. Thus, although the micromotion measurements are an overestimation of the true cup micromotion, they still are comparable to one another for each loading configuration and demonstrate the effect of considering dynamic hip motion on the levels of micromotion. Both the loading cycles and the flexion-extension cycles were modelled as sine waves as they are a close approximation of the true loading and motion profiles and have been used in many studies [2,7,10–12,36,37]. This study only looked at hip flexion-extension as it is the predominant direction of motion in both walking and Stair Climbing [28], and hence was most likely to have an effect on cup micromotion.

A limitation of this study was the use of a synthetic acetabular model rather than cadaveric bones. However, the similarity in cup micromotion under static conditions using this model and those published in the literature using cadaveric pelvic bones emphasise the appropriateness of this model for micromotion studies.

Conclusions

A six DoF measurement system and acetabular model presented in a previous study were combined with a dynamic hip motion simulator to create a novel pre-clinical testing tool to investigate the effect of dynamic hip motion and two different porous coatings on the micromotion of a press-fit cup. The results of this study demonstrated a statistically significant increase in micromotion when the cups were subjected to dynamic hip motions compared to Static Flexion. This study has clearly shown that in order to investigate the stability of acetabular cups *in vitro*, *in vivo* post-op conditions need to be replicated, including cyclic loading and dynamic hip motion. This new method offers a unique pre-clinical tool to assess the performance of an acetabular cup and identify features that influence cup fixation and stability, such as design features, acetabular defects and surgical techniques, amongst others.

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Declaration of conflicts of interest

The authors declare that there is no conflict of interest.

Ethical approval

Not applicable

References

[1] Konttinen YT, Zhao D, Beklen A, Ma G, Takagi M, Kivelä-Rajamäki M, et al. The microenvironment around total hip replacement prostheses. Clin Orthop Relat Res 2005:28–38.

[2] Won C, Hearn T, Tile M. Micromotion of cementless hemispherical acetabular components. Does press-fit need adjunctive screw fixation? J Bone Jt Surg Br 1995;77-B:484–9.

[3] Jasty M, Bragdon C, Burke D, O'Connor D, Lowenstein J, Harris WH. In vivo skeletal responses to porous-surfaced implants subjected to small induced motions. J Bone Joint Surg Am 1997;79:707–14.

[4] Pilliar RM, Lee JM, Maniatopoulos C. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. Clin Orthop Relat Res 1986:108–13.

[5] Hori RY, Lewis JL. Mechanical properties of the fibrous tissue found at the bone-cement interface following total joint replacement. J Biomed Mater Res 1982;16:911–27.

[6] Fritsche A, Bialek K, Mittelmeier W, Simnacher M, Fethke K, Wree A, et al. Experimental

investigations of the insertion and deformation behavior of press-fit and threaded acetabular cups for total hip replacement. J Orthop Sci 2008;13:240–7.

[7] Stiehl JB, MacMillan E, Skrade DA. Mechanical stability of porous-coated acetabular components in total hip arthroplasty. J Arthroplasty 1991;6:295–300.

[8] Perona PG, Lawrence J, Paprosky WG, Patwardhan AG, Sartori M. Acetabular micromotion as a measure of initial implant stability in primary hip arthroplasty. An in vitro comparison of different methods of initial acetabular component fixation. J Arthroplasty 1992;7:537–47.

[9] Kwong LM, O'Connor DO, Sedlacek RC, Krushell RJ, Maloney WJ, Harris WH. A quantitative in vitro assessment of fit and screw fixation on the stability of a cementless hemispherical acetabular component. J Arthroplasty 1994;9:163–70.

[10] Pitto RP, Willmann G, Schramm M. [Initial Stability of Modular Acetabular Components. Comparative In-vitro Study with Polyethylene and Ceramic Liners]. Biomed Tech Eng 2001;46:109–12.

[11] von Schulze-Pellengahr C, Bürkner A, Lichtinger T, Teske W, Fottner A, Wegener B, et al. [Does osteoporosis lead to reduction the primary stability of cementless hip cups?]. Orthopade 2011;40:607–13.

[12] Crosnier EA, Keogh PS, Miles AW. A novel method to assess primary stability of press-fit acetabular cups. Proc Inst Mech Eng H 2014;228:1126–34.

[13] Stryker. Stryker Tritanium acetabular shell n.d. http://www.stryker.com/en-us/products/Orthopaedics/HipReplacement/Acetabular/TritaniumAcetabularShell/index.htm (accessed January 26, 2015).

[14] Cowin SC. Bone mechanics handbook. 2nd ed. Boca Raton, FL: CRC Press; 2001.

[15] Helgason B, Perilli E, Schileo E, Taddei F, Brynjólfsson S, Viceconti M. Mathematical relationships between bone density and mechanical properties: a literature review. Clin Biomech (Bristol, Avon) 2008;23:135–46.

[16] Baleani M, Fognani R, Toni A. Initial Stability of a Cementless Acetabular Cup Design: Experimental Investigation on the Effect of Adding Fins to the Rim of the Cup. Artif Organs 2001;25:664–9.

[17] Macdonald W, Carlsson L V, Charnley GJ, Jacobsson CM. Press-fit acetabular cup fixation: principles and testing. Proc Inst Mech Eng Part H J Eng Med 1999;213:33–9.

[18] Adler E, Stuchin SA, Kummer FJ. Stability of press-fit acetabular cups. J Arthroplasty 1992;7:295–301.

[19] Small SR, Berend ME, Howard LA, Rogge RD, Buckley CA, Ritter MA. High initial stability in porous titanium acetabular cups: a biomechanical study. J Arthroplasty 2013;28:510–6.

[20] Schmidig G, Patel A, Liepins I, Thakore M, Markel DC. The effects of acetabular shell deformation and liner thickness on frictional torque in ultrahigh-molecular-weight polyethylene acetabular bearings. J Arthroplasty 2010;25:644–53.

[21] Jin ZM, Meakins S, Morlock MM, Parsons P, Hardaker C, Flett M, et al. Deformation of press-fitted metallic resurfacing cups. Part 1: experimental simulation. Proc Inst Mech Eng Part H J Eng Med 2006;220:299–309.

[22] Meding JB, Small SR, Jones ME, Berend ME, Ritter MA. Acetabular cup design influences deformational response in total hip arthroplasty. Clin Orthop Relat Res 2013;471:403–9.

[23] Haeussler K, Kruse C, Flohr M, Preuss R, Streicher R, Morlock M. Stress Analysis of Ceramic Acetabular Liners Under in Vivo Like Loading Conditions. ISTA Annu. Conf., Kyoto: 2014.

[24] Bishop N, Morlock M. The risk of acetabular cup turn-out due to friction moments. J Biomech

2008;41:S326.

[25] Michel A, Bosc R, Vayron R, Haiat G. In vitro evaluation of the acetabular cup primary stability by impact analysis. J Biomech Eng 2015;137:031011. doi:10.1115/1.4029505.

[26] Riener R, Rabuffetti M, Frigo C. Stair ascent and descent at different inclinations. Gait Posture 2002;15:32–44.

[27] Taylor SJG, Perry JS, Meswania JM, Donaldson N, Walker PS, Cannon SR. Telemetry of forces from proximal femoral replacements and relevance to fixation. J Biomech 1997;30:225–34.

[28] Andriacchi TP, Andersson GB, Fermier RW, Stern D, Galante JO. A study of lower-limb mechanics during stair-climbing. J Bone Joint Surg Am 1980;62:749–57.

[29] Bergmann G, Deuretzbacher G, Heller M, Graichen F, Rohlmann A, Strauss J, et al. Hip contact forces and gait patterns from routine activities. J Biomech 2001;34:859–71.

[30] Murray MP, Kory RC, Clarkson BH. Walking patterns in healthy old men. J Gerontol 1969;24:169–78.

[31] Paul JP. Force Actions Transmitted by Joints in the Human Body. Proc R Soc B Biol Sci 1976;192:163–72.

[32] Nadeau S, McFadyen B., Malouin F. Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: what are the challenges compared to level walking? Clin Biomech 2003;18:950–9.

[33] Bergmann G, Graichen F, Rohlmann A. Is staircase walking a risk for the fixation of hip implants? J Biomech 1995;28:535–53.

[34] Bergmann G, Graichen F, Rohlmann A, Bender A, Heinlein B, Duda GN, et al. Realistic loads for testing hip implants. Biomed Mater Eng 2010;20:65–75.

[35] Zivkovic I, Gonzalez M, Amirouche F. The effect of under-reaming on the cup/bone interface of a press fit hip replacement. J Biomech Eng 2010;132:041008.

[36] Clements JP. Design and development of pre-clinical hip stem stability testing methods. PhD, University of Bath, 2006.

[37] Liu C, Green SM, Watkins ND, Gregg PJ, McCaskie AW. A preliminary hip joint simulator study of the migration of a cemented femoral stem. Proc Inst Mech Eng Part H J Eng Med 2003;217:127–35.

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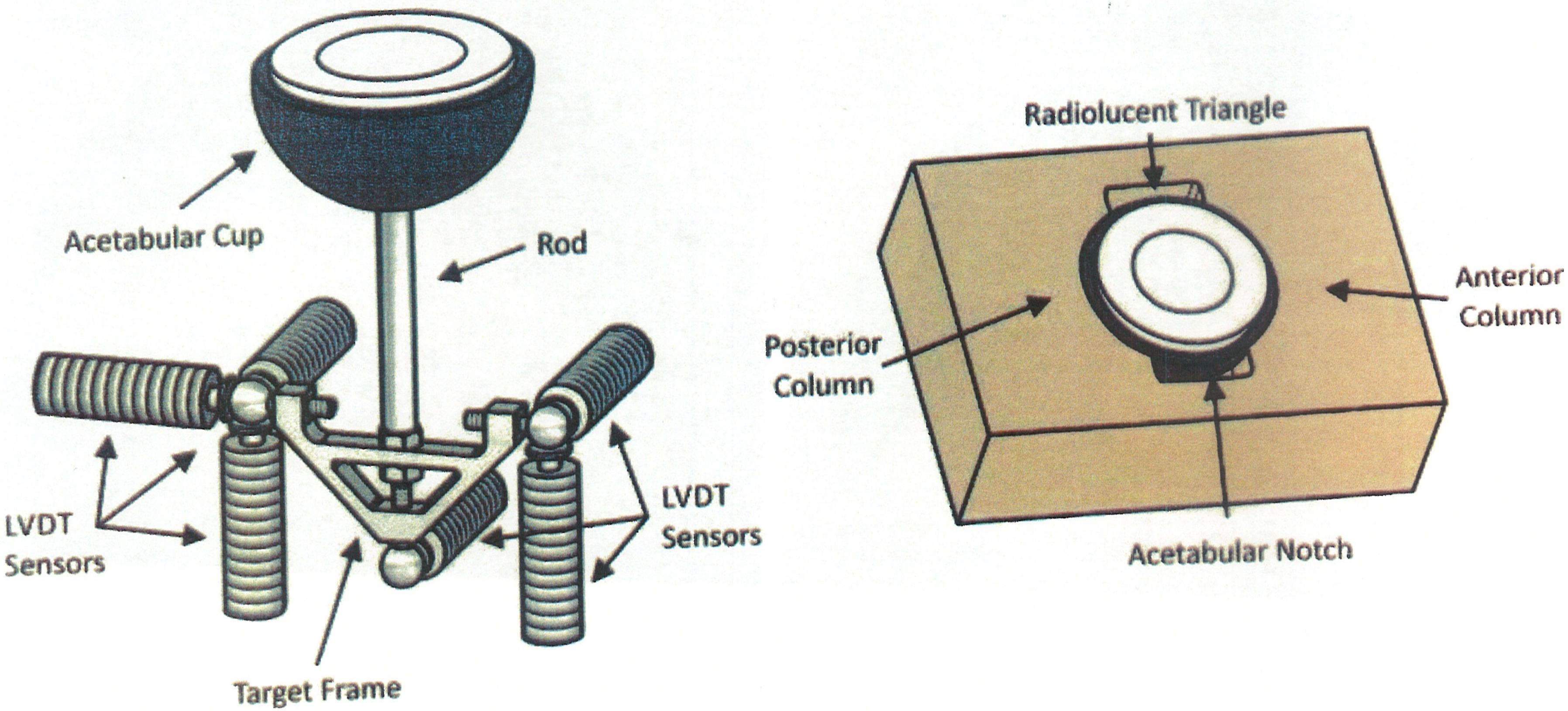

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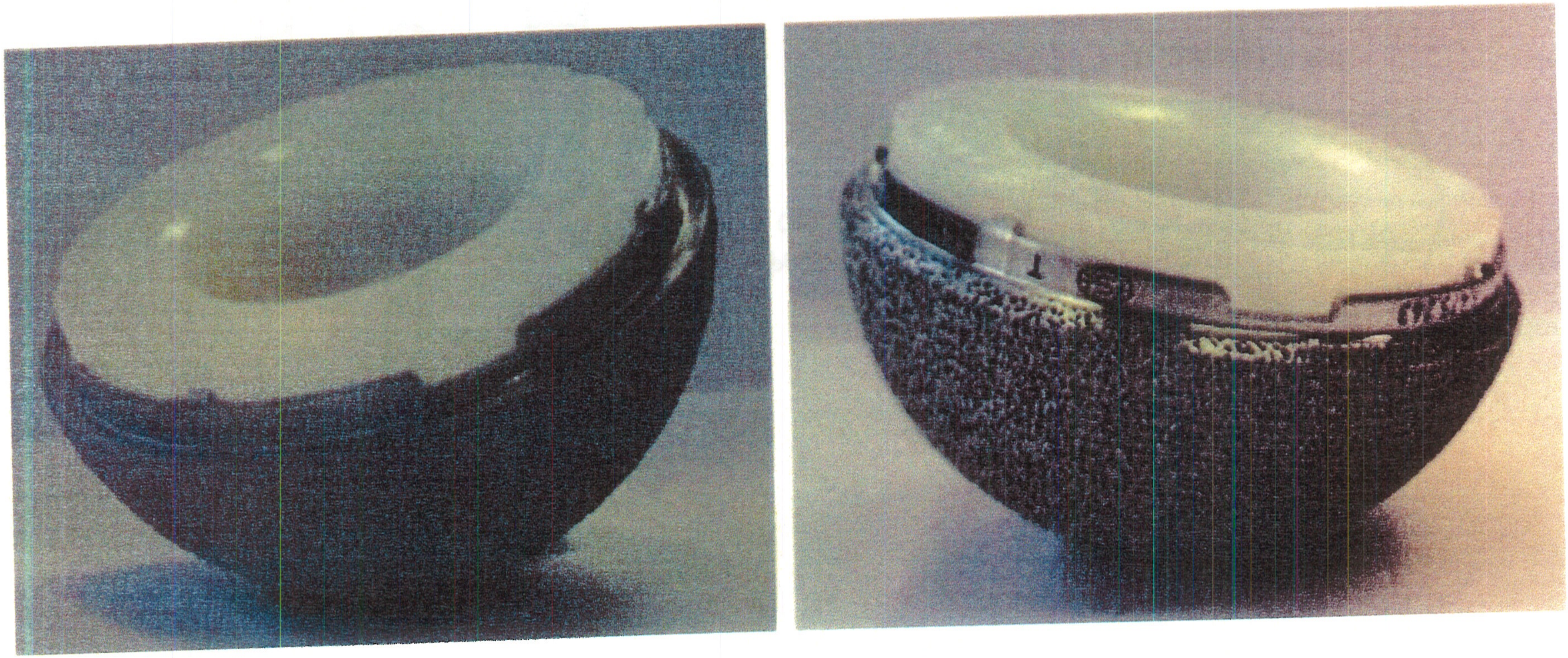

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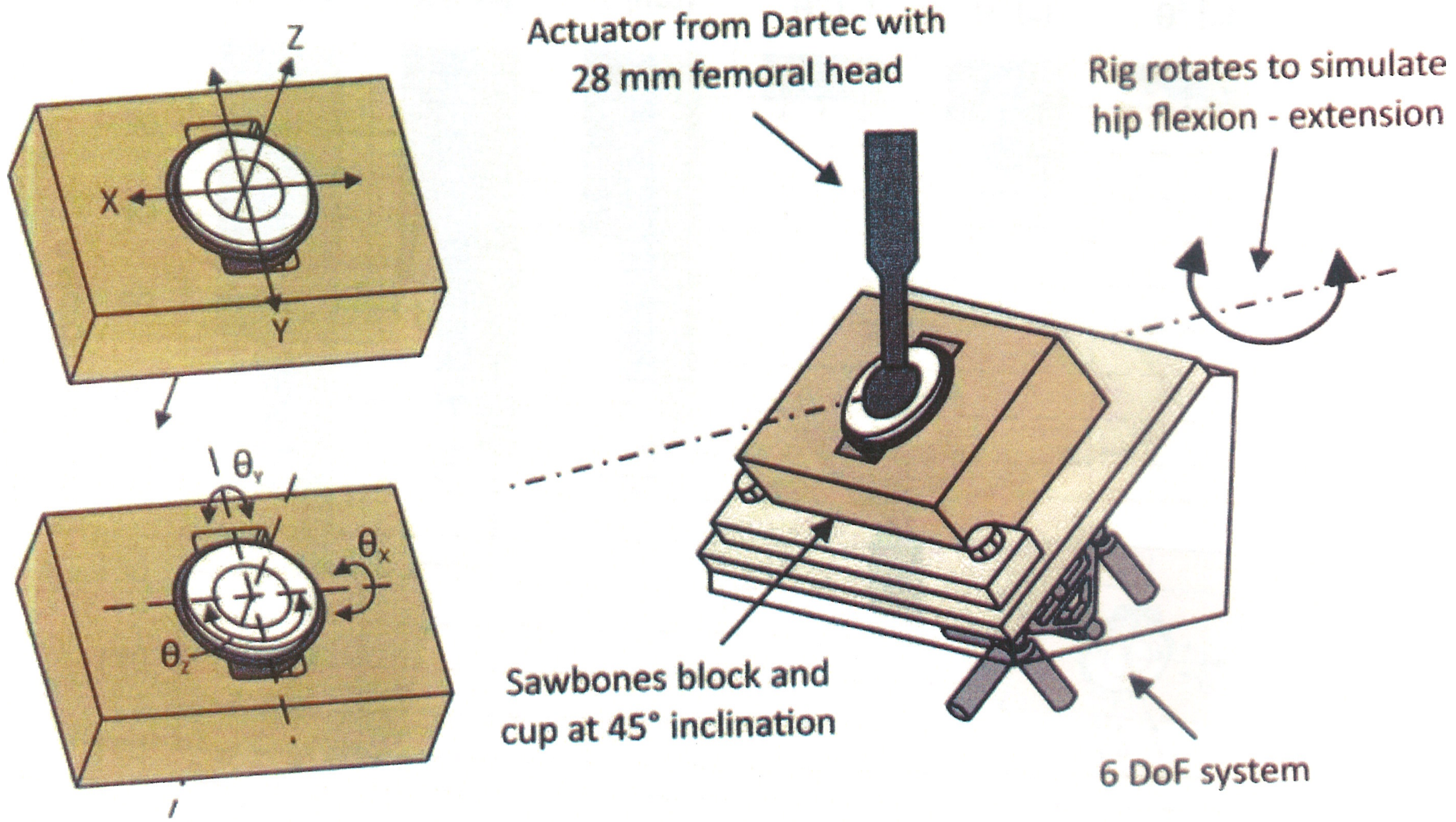

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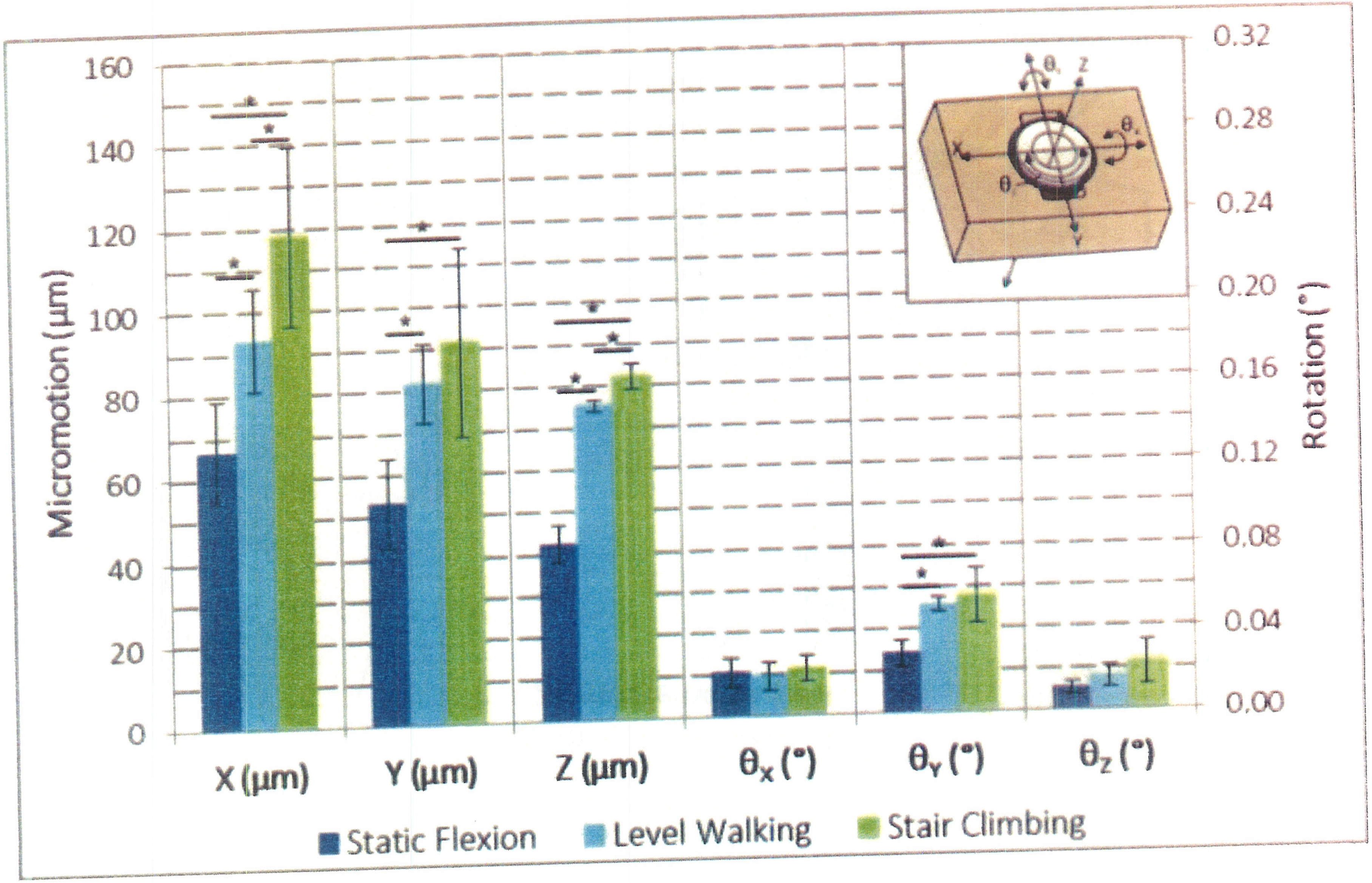

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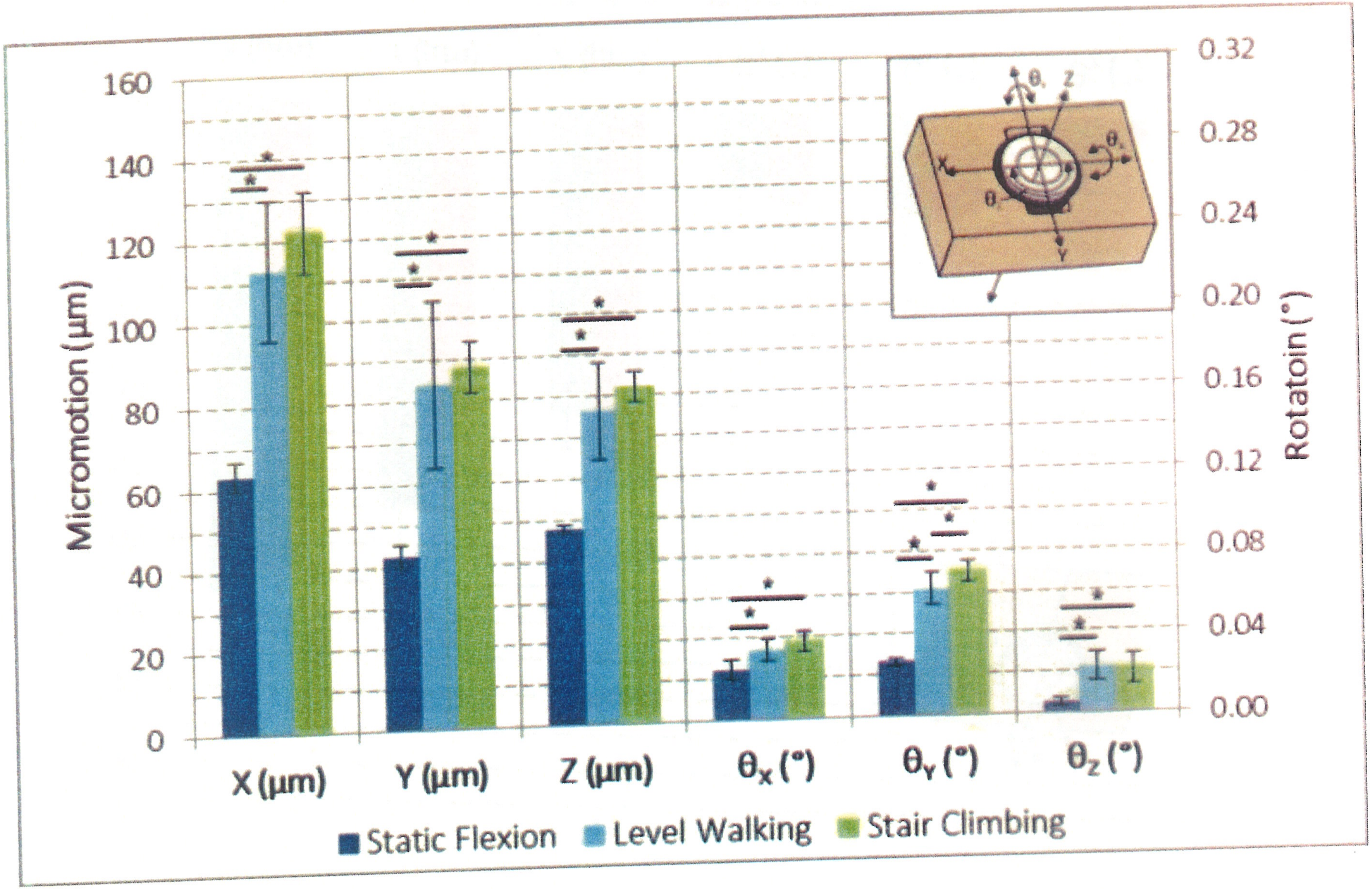

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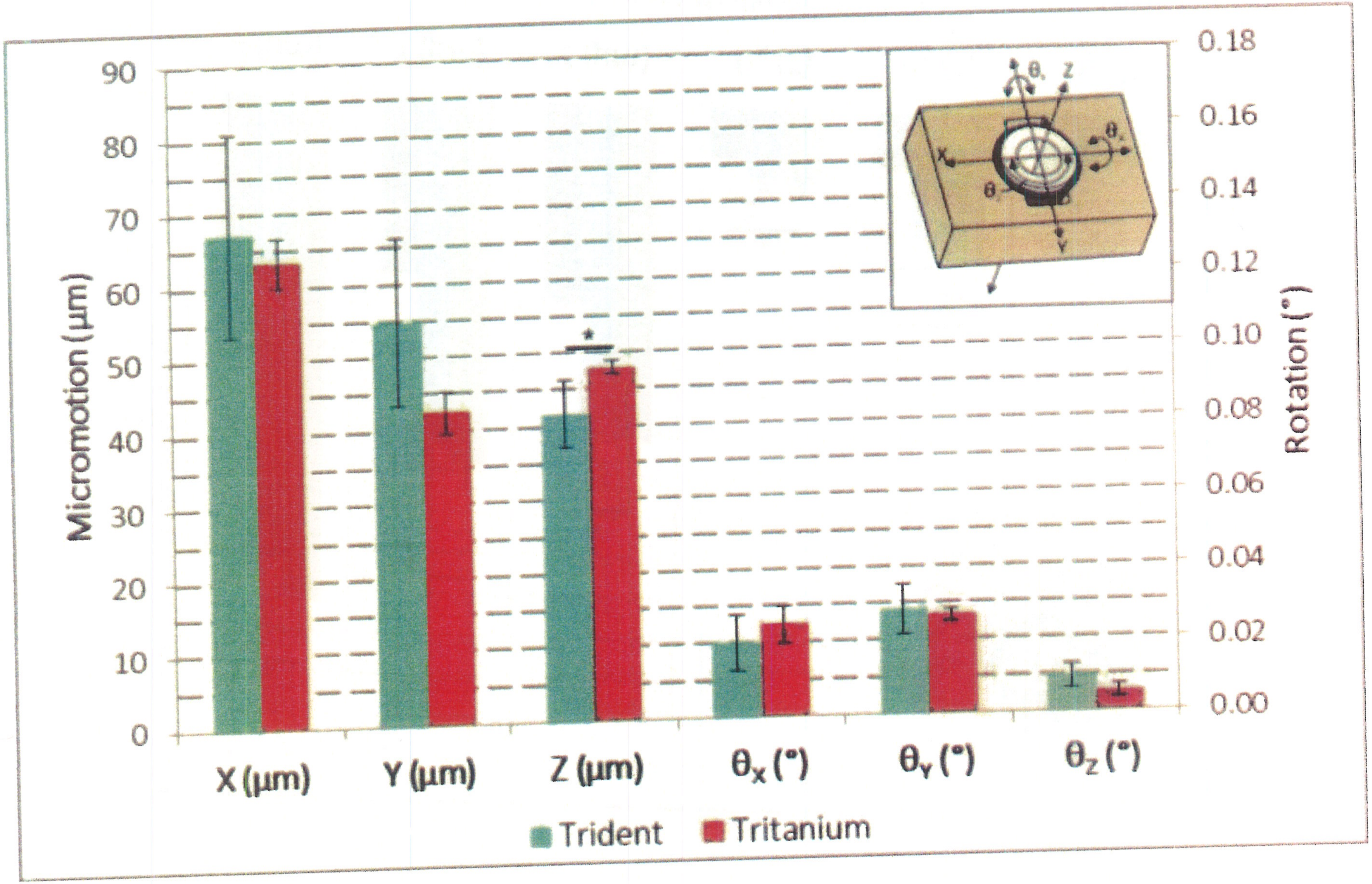

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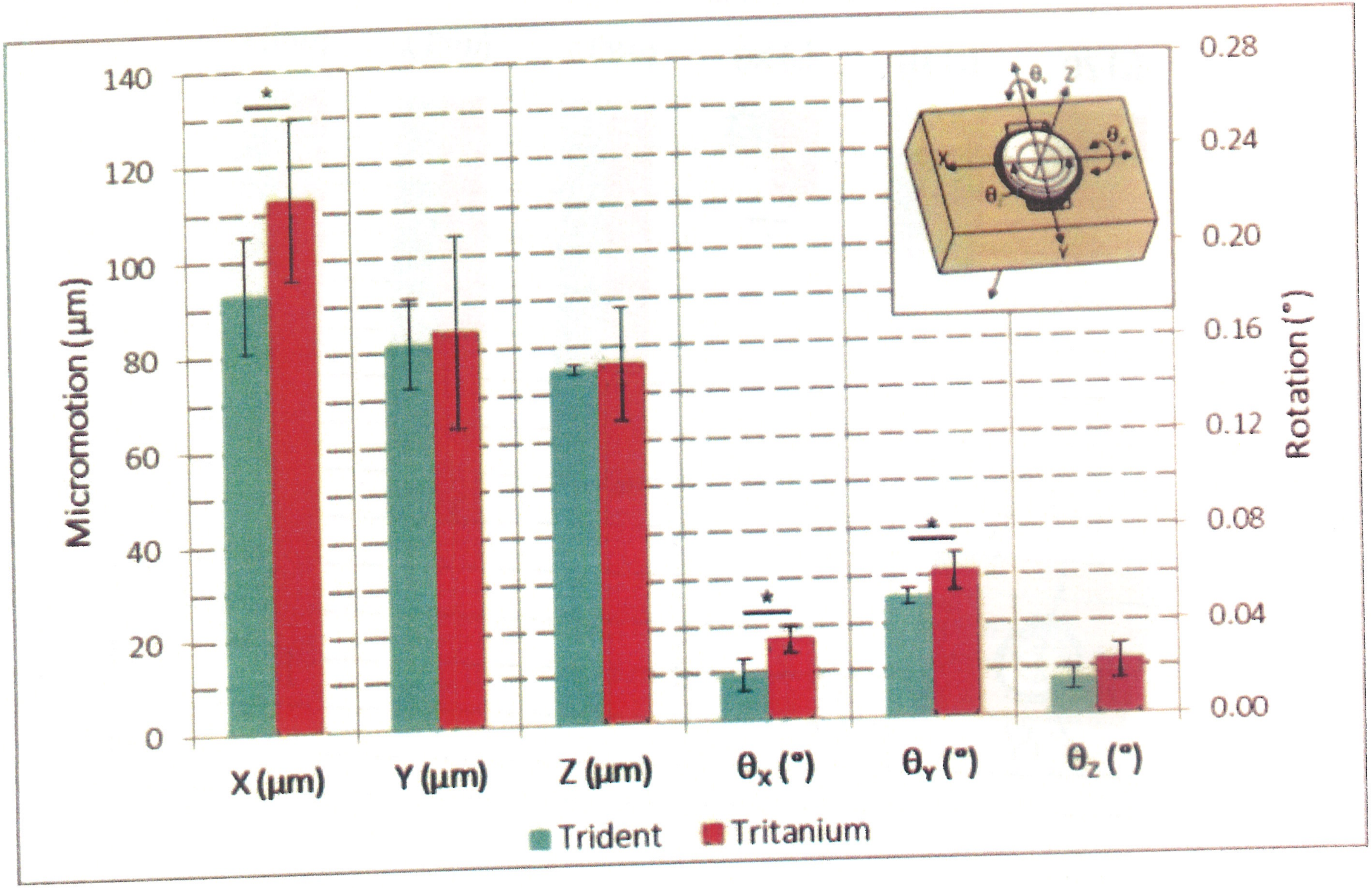

Figure 8 (colour)
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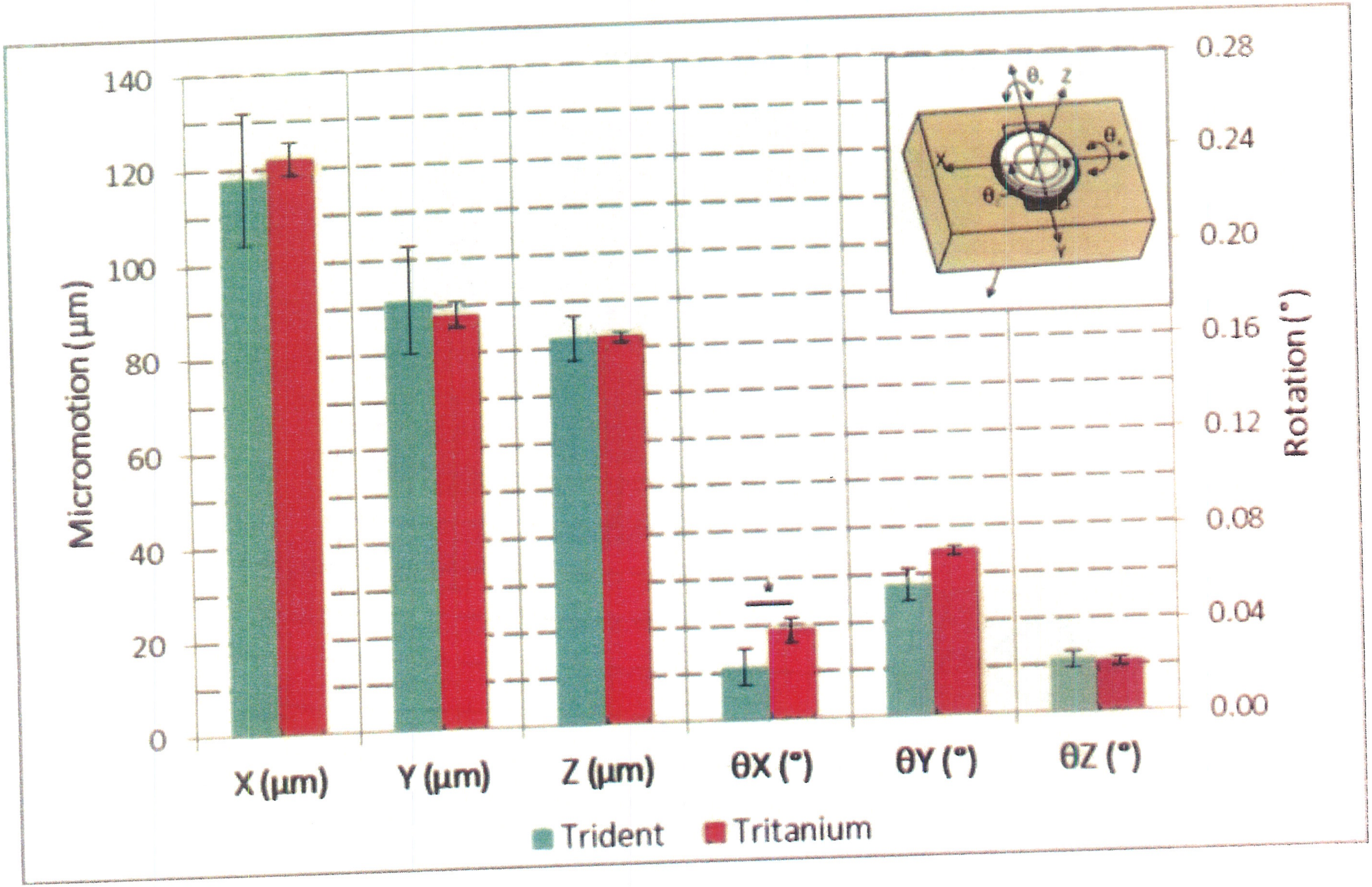

Figure 9 (colour)
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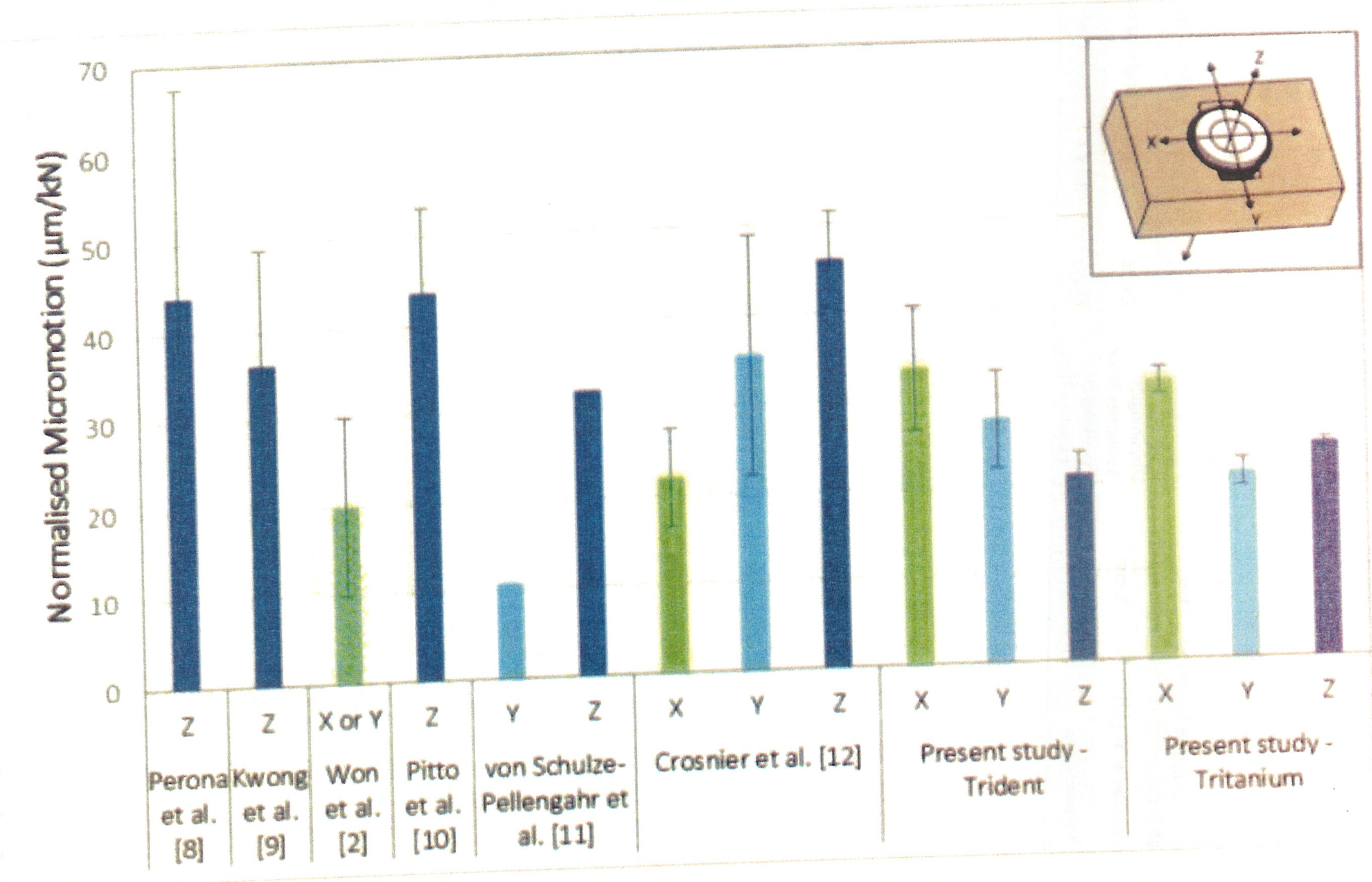

Table 1 (colour)

Study	Press-fit	Acetabular model	Loading protocol	Peak load	Hip / Cup Orientation
Perona et al. [8]	2 mm	Fresh-frozen	Ramp	2.4 kN	Anatomical position
Kwong et al. [9]	1 mm	Embalmed	Ramp	1.1 kN	Single leg stance
Won et al. [2]	1 mm	Fresh-frozen	Cyclic	1.5 kN	30° inclination & 20° anteversion
Pitto et al. [10]	2 mm	Sawbones Hemipelvis	Cyclic	2.4 kN	60° to horizontal
von Schulze-Pellengahr et al. [11]	Not specified	Macerated	Cyclic	3.8 kN	61° to vertical
Crosnier et al. [12]	1 mm	Physiological Sawbones Block	Cyclic	2.0 kN	Single leg stance
Present Study	1 mm	Physiological Sawbones Block	Cyclic	2.0 kN	Heel Strike